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INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(51) International Patent Classification [6] : **A61B 8/00**	**A1**	**(11) International Publication Number:** **WO 96/28095**
		(43) International Publication Date: 19 September 1996 (19.09.96)

(54) Title: IMAGING SYSTEM DISPLAY PROCESSOR

(57) Abstract

A display processor (10) for an ultrasonic imaging system includes a two-dimensional filter (12) to generate a smoothed image signal I_P from a high spatial resolution image signal I_D. I_P is optimized for high contrast resolution and good tissue differentiation, and I_D is optimized for high detail resolution and the display of fine structural details. I_P and I_D are applied as addressing inputs to a look-up table (14) that provides an output image signal I_O that combines both detail resolution of I_D and contrast resolution of I_P. I_O can display detail resolution as brightness and contrast resolution as color. I_O can also be formed as a weighted combination or sum of I_P and I_D.

IMAGING SYSTEM DISPLAY PROCESSOR

BACKGROUND OF THE INVENTION

This invention relates to imaging systems, and in particular to a display processor that enhances the simultaneous display of multiple types of resolution in such systems.

Ultrasound images are used diagnostically for two different purposes: the display of structural details of tissue, and the differentiation of different types of tissue. A high spatial resolution (or detail resolution) in the image maximizes the ability of the user to detect fine structural details. A high contrast resolution in the image maximizes the user's ability to detect tissue type differences.

One of the fundamental limitations on contrast resolution is speckle noise. Speckle noise is inherent to coherent imaging systems and is caused by coherent (phase-sensitive) interference of waves scattered by structures too fine to be resolved. Speckle noise does not carry information on the structural details of the object, but its low frequency components carry information on the mean backscattering strength of the object. For this reason, high frequency components of speckle noise should be reduced in order to improve contrast resolution.

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Various signal processing techniques have
been used to improve contrast resolution. Low-pass
video filters, low-pass scan conversion filters, and
even the limited bandwidth of the display monitor can
be used to reduce high-frequency spatial variations of
speckle noise. In this way, differences in mean
backscattering strength from one tissue to another can
be significantly accentuated. The spatial smoothing
associated with this approach to improving contrast
resolution significantly reduces detail resolution.

Bamber U.S. Patent 4,783,839 discloses
a system for reducing speckle noise that varies the
amount of smoothing across the image as a function of
how closely individual regions of the image resemble
speckle. In many cases speckle noise is superimposed
on resolvable structural details of the object, and
diagnostically useful detail resolution therefore can
be lost.

Non-linear display mapping is another signal
processing technique that has been used to improve
contrast resolution. Especially when cascaded with
smoothing video filters, non-linear display maps can
increase contrast resolution at selected display
intensity levels at the expense of reduced detail and
contrast resolution at all other display intensity
levels.

Speckle noise can also be reduced during
image formation by any of several incoherent averaging
(compounding) techniques, including spatial, frequency,
and temporal compounding.

Spatial compounding during image formation
can be accomplished by shifting the transducer in the
lateral, axial, or elevational direction. Frequency
compounding is performed by dividing the passband of
input pulses into several sub-bands in the frequency
domain. Temporal compounding produces images by
incoherently summing range intensity samples.

These compounding techniques exploit the fact
that the ensemble average of a speckle image is the
same as the incoherent average of the original object.
Since the mottled appearance of speckle noise carries
information only about the imaging device and not the
imaged object, speckle variations can be reduced by
incoherently averaging independent measurements without
altering original target contrast. Reduced speckle
variation results in improved contrast resolution at
the price of reduced detail resolution.

Lipschultz U.S. Patent 5,224,483 discloses
a system for enhancing an ultrasound image by reducing
clutter in a blood pool area of the image. The blood
pool areas are identified by means of low-pass
filtering and non-linear intensity mapping, and a mask
signal is generated having substantially a first value
in areas of tissue and substantially a second value in
areas of blood pool. The image signal is then
modulated with the mask signal to substantially remove
clutter in the blood pool. The mask signal preferably
is not strictly binary, but contains some intermediate
levels to provide a smooth transition between masked
and unmasked regions, so as to prevent an unnatural
appearance of the final image. This technique only
suppresses clutter in blood pool areas, and does not
address the problem of improved contrast resolution in
tissue areas.

A need exists for an improved image processor
that enhances contrast resolution to improve
differentiability of tissue types (liver versus kidney,
healthy tissue versus lesion) without losing
diagnostically important fine structural details of the
image.

SUMMARY OF THE INVENTION

According to this invention, an imaging
system display processor is responsive to first and

second image signals depicting a common entity. The
first image signal has greater detail resolution than
the second image signal, while the second image signal
has greater contrast resolution than the first image
signal. The display processor forms an output signal
as a function of the two image signals such that the
output signal combines detail information of the first
image signal and contrast information of the second
image signal.

The display processor preferably comprises a
display generator such as a two-dimensional map. In
some embodiments the output signal is characterized by
an intensity which displays detail resolution of the
first image signal and a color which displays contrast
resolution of the second image signal. In other
embodiments the output signal is formed as a weighted
sum of the first and second image signals, and the
weighted sum comprises a weighing factor that varies in
accordance with both of the first and second image
signals.

BRIEF DESCRIPTION OF THE DRAWINGS

Figure 1 is a block diagram of an ultrasonic
imaging system that incorporates a preferred embodiment
of this invention.

Figure 2 is a block diagram of the filter of
Figure 1.

Figure 3 is a block diagram of the azimuthal
filter of Figure 2.

Figure 4 is a block diagram of the range
filter of Figure 2.

Figure 5 is a graph relating to a first
preferred embodiment of the display generator of
Figure 1.

Figure 6 is a graph of $\alpha(I_P, I_D)$ that relates
to a second preferred embodiment of the display
generator of Figure 1.

Figure 7 is a graph of $F'(I_P, I_D)$ that
relates to the second preferred embodiment of the
display generator of Figure 1.

Figure 8 is a block diagram of an ultrasonic
imaging system that incorporates another embodiment of
this invention.

DETAILED DESCRIPTION OF THE
PRESENTLY PREFERRED EMBODIMENTS

Turning now to the drawings, Figure 1 is
a block diagram of a first preferred embodiment of this
invention, which responds to a high detail resolution
image signal I_D generated by any suitable ultrasonic
imaging system S. The system S is conventional, and in
this embodiment the signal I_D is a standard B-mode
image signal that varies in intensity as a function of
the backscattering coefficient of the tissue. Of
course, this invention is not limited to use with
B-mode image signals, and ultrasonic imaging systems
that monitor any other suitable acoustic properties of
tissue (including but not limited to attenuation
coefficient, speed of sound, and elasticity) can also
be used with this invention. The image signal I_D can
be an unprocessed image signal, or an image signal with
some amount of video filtering (either smoothed or edge
enhanced).

As shown in Figure 1, the image signal I_D is
applied as an input to a display processor 10 that
includes a two dimensional filter 12 and a display
generator 14 such as a two dimensional lookup table.
The two dimensional filter 12 can be a video
filter which generates as an output a processed, high
contrast resolution image signal I_P. The image signal
I_P has lower detail resolution and higher contrast
resolution than the image signal I_D.

The degree of smoothing provided by the
filter 12 is selected to enhance contrast resolution
and therefore differentiability of different tissue

types. Generally, the image signal I_P is a multi-level
signal that varies across tissues of varying types and
in which major regions of differing types of solid
tissue can be distinguished from one another. For
example, the image signal I_P for a region including
blood pool and at least first and second types of solid
tissue preferably allows the blood pool to be
distinguished from the solid tissue, and the first and
second types of solid tissue to be distinguished from
each other. The size and shape of the two-dimensional
spatial filter 12 can vary widely, as suitable for the
application and the body type of the patient. Multiple
user selectable filters can be provided, or a single
filter can be user-adjustable.

Many implementations for the two dimensional
spatial filter 12 are known to those skilled in the
art. One suitable approach is shown in Figures 2-4,
and includes an image memory 24, an azimuthal
filter 26, a range filter 28, and a scaling circuit 30.
The image memory 24 stores lines of acoustic data
obtained from the detail image signal I_D, the range
filter 28 filters along the lines, and the azimuthal
filter 26 filters between lines.

As shown in Figure 3, the azimuthal filter 26
includes 2 FIFO registers 32, 34, which both receive
the same image signal from the memory 24. The
registers supply output signals to a summer 36 under
control of a clock signal. A delay element 38 shifts
the output signal of the register 34 in time by an
amount D_1 equal to a desired number of lines. Assuming
the original image signal is $I_D(l_n, t_n)$, where l_n is
the line number and t_n is the range sample number, then
the output I_S of the summer 36 can be expressed as
follows:

$$I_S(l_n, t_n) = I_D(l_n, t_n) - I_D(L_n - D_1, t_n).$$

The signal I_S is then applied to an
accumulator 40, which includes a summer 42 and a line

FIFO register 44. The accumulator 40 integrates the
line data and generates an output signal I_{INT}:

$$I_{INT}(l_n, t_n) = \sum_{l_n' = l_n}^{l_n + D_1 - 1} I_D(l_n', t_n).$$

I_{INT} is therefore the sum of D_1 adjacent lines. This
corresponds to a box-car filter of size D_1.

Figure 4 provides details of the range
filter 28, which calculates difference and accumulates
as a function of range sample number rather than line
number. Otherwise, the range filter 28 functions in a
similar manner to the azimuthal filter 26 discussed
above.

The size of the two-dimensional filter 12 can
be selected between upper and lower limits. The upper
limit is equal to the size of the smallest lesion or
other tissue type to be detected. The lower limit is
affected by the average speckle size and the desired
improvement in contrast to noise ratio (CNR). Average
speckle size is equal to the area (e.g., at -6.8dB) of
the autocorrelation of the point spread function. CNR
is defined as follows:

$$CNR = \frac{\Delta I}{(\sigma_\beta^2 + \sigma_l^2)^{1/2}}$$

ΔI is the average intensity (backscattering
coefficient) difference between the lesion and
background, and σ_β^2 and σ_l^2 are the variances of the
background and the lesion, respectively. If CNR is to
be increased by 3dB, filter size should be set equal to
the average speckle size.

In embodiments such as that of Figure 1,
where I_P is generated from I_D, the high detail
resolution of I_D benefits the contrast resolution of

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I_P. This is because a higher detail resolution brings with it more independent samples that are available for processing (averaging). The improvement is proportional to the square root of the number of independent samples. As explained above, high detail resolution is not a sufficient condition for high contrast resolution, since high spatial frequency components of the image signal I_D can have an adverse effect on a user's ability to perceive contrast differences in the image.

It is not essential that the high contrast resolution image signal I_P must be obtained from the image signal I_D with a video filter. Other techniques such as other types of spatial filtering, frequency compounding, temporal compounding, spatial compounding, combinations of these techniques, and other speckle reduction techniques can be used. In some cases it may be preferable to use beam formation techniques to generate the image signals I_P and I_D directly. In this case there will often be no need to include any type of two dimensional filter in the image processor. Generally, it is simpler to process the same beam-formed image signal twice (once for detail resolution to create I_D and once for contrast resolution to create I_P).

Once I_D and I_P have been formed as described above, they are then applied to first and second inputs 16, 18 of the display generator 14, which generates an output image signal I_O. I_O combines or superimposes the detail information of I_D with the contrast information of I_P so that superimposed detail and contrast information are both preserved, and the human eye can distinguish both when the signal I_O is displayed on a monitor 20. At higher spatial resolutions, I_O varies in accordance with I_D. At lower spatial resolutions I_O varies in accordance with I_P.

Thus, the output image signal I_O is
a composite of the signal I_D, which is acquired and
processed to maximize detail resolution, and the signal
I_P, which is acquired and processed to maximize
contrast resolution.

In this embodiment, the display generator 14
is a two dimensional look-up table in which I_D and I_P
are used as addresses to select the corresponding
stored value I_O from the table. I_D and I_P are
synchronized such that at any given point in time they
relate to the same physical region of the object being
imaged.

The look-up table of the display generator 14
in this embodiment is a color table that assigns
a pixel value to I_O based on both I_D and I_P as in
equation (1):

$$I_O = F(I_P, I_D) \qquad\qquad (1)$$

The function F in equation (1) can be thought of as
a vector function, with each component of the vector
function corresponding to a display color. When
a conventional RGB (red-green-blue) monitor 20 is used,
F is preferably a three-dimensional vector function.
When the red, green and blue components are set to be
equal to one another, the corresponding value of I_O
will be a gray scale value. The exact shape of F can
be set to emphasize various aspects of image quality.
For example, F can be set to reduce noise by having
a low value of F when I_P is low, even if I_D is high.
Contrast resolution and detail resolution are both
preserved in I_P by setting F such that the brightness
of a particular pixel of I_O varies with I_D while the
color of this pixel varies with I_P. When this is done,
two different tissues which differ in contrast levels
will be displayed in different colors. Figure 5 shows
one possible shape for the three components of a three-

dimensional F for a given value of I_D and varying
values of I_p.

The following equations 2-5 can be used to
determine red, green, and blue components of I_O as a
function of I_D and I_p:

$$I_O = I_D{}' \, K_{R,G,B}(I_p) ; \qquad\qquad (2)$$

$$I_D{}' = \gamma I_D + (1-\gamma) I_p ; \qquad\qquad (3)$$

$$\gamma = \frac{1-\beta}{1 + \left(\dfrac{|I_P - I_D|}{\tan(.5\pi(1-\alpha))}\right)^6} ; \qquad\qquad (4)$$

$$K_{R,G,B}(I_P) = L_{R,G,B} + \frac{(H_{R,G,B} - L_{R,G,B})}{(1 + \exp(-4S_{R,G,B}(I_P - B_{R,G,B})))} . \quad (5)$$

The following constants can be used in Eq. 2-5, where
subscripts R,G,B are used for the red, green and blue
components, respectively:

$$L_R = .5, \quad L_G = .8, \quad L_B = .6 ;$$
$$H_R = 1, \quad H_G = .9, \quad H_B = .9 ;$$
$$S_R = 2, \quad S_G = 2, \quad S_B = 2 ;$$
$$B_R = .2, \quad B_G = .2, \quad B_B = .2 ;$$
$$\alpha = .7, \quad \beta = .1.$$

With this approach, the image signal I_D with
optimized detail resolution is superimposed on the
image signal I_p with optimized contrast resolution by
color shading. In this way enhanced contrast
information and enhanced detail information are
presented simultaneously via the signal I_O on the
monitor 20, and both types of information can be
perceived by the user.

Figures 5 and 6 relate to a second preferred embodiment of the display generator 14. In this case the display generator generates the output image signal I_O as a monochromatic, two-dimensional mapping of I_D and I_P:

$$I_O = F'(I_P, I_D). \qquad\qquad (6)$$

The explicit form of $F'(I_P, I_D)$ is given by equation (7):

$$I_O = I_P + \alpha(I_P, I_D) \bullet (I_D - I_P), \qquad (7)$$

where $\alpha(I_P, I_D)$ is a weighing factor which is a function of I_P and I_D. I_O is thus a weighted combination of I_P and I_D. For example, when $\alpha = 0$, $I_O = I_P$, and when $\alpha = 1$, $I_O = I_D$. By varying α between 0 and 1, the output image signal I_O can be made equal to any weighted combination of I_P and I_D between these two extremes.

The mean brightness of the output image I_O is given by equation (8), where <> denotes a spatial averaging operation:

$$<I_O> = <I_P> + <\alpha \bullet (I_D - I_P)> \qquad (8)$$

For practical purposes, the second term of equation (8) will be zero or nearly zero. Hence, the mean brightness of the image signal I_O will be approximately equal to that of I_P.

The function α can be designed based on known characteristics of ultrasound images. For example, noise in a dark background can be removed or substantially reduced by setting α close to zero when I_P is low. In order to avoid the undesirable suppression of small objects in a dark background, α can be given a larger value when the difference between I_P and I_D is small.

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As long as the size of a small object is larger than
the size of noise speckles, or the object is brighter
than the noise, the value of $(I_D - I_P)$ tends to be
smaller for the object than the noise. For this reason
α is preferably set to zero when I_P is low and $(I_D -
I_P)$ is large. Similarly, when the intensity of the
background is high (I_P is large) and $(I_P - I_D)$ is
large, α can be set close to zero to suppress dark
holes in a bright background, reduce speckle noise, and
achieve better contrast resolution in relatively bright
tissue areas.

When I_D and I_P are nearly equal, I_D
represents mainly image details, especially when I_P
ranges from middle to high levels. In this case α is
set to a high value to preserve these image details.

Figure 6 illustrates a preferred form of
$\alpha(I_P, I_D)$ which generally follows the considerations
set out above. A good balance between noise suppres-
sion and detail preservation can be achieved by
adjusting the width of the diagonally-extending ridge
in Figure 6 and varying the height of the ridge along
the diagonal.

The display generator 14 can be implemented
as a two-dimensional look-up table having I_P and I_D as
input addresses, because $\alpha(I_P, I_D)$ and consequently
$F'(I_P, I_D)$ can be pre-calculated and stored in the
table. Figure 7 shows $I_O = F'(I_P, I_D)$, calculated
using $\alpha(I_P, I_D)$ as shown in Figure 6. Of course,
several different look-up tables may be stored for use
with particular applications or types of tissue or
patient.

The approach discussed above in connection
with Figures 6 and 7 provides the important advantage
that it can suppress noise in dark areas (e.g. clutter
in a blood pool) and reduce speckle noise in light
tissue areas, with little or no loss of detail of small

structures and edges.   In addition, the mean brightness
of the image is preserved.

The two-dimensional filter 12 of the display
processor 10 is not required in all embodiments.   The
processed image signal I_p can be generated in any
suitable manner to provide valid contrast information
between tissues of different types.

In some applications, it is highly desirable
to suppress blood pool in the displayed image in order
to enhance contrast and to emphasize boundaries.
Motion information can readily be obtained as described
below by measuring temporal coherence, or correlation
between subsequent frames.   This motion information can
be used to detect blood pool and to suppress blood pool
images without affecting tissue detail information.

Figure 8 shows one such embodiment, which
substitutes a motion estimator 22 for the two-
dimensional filter 12 of Figure 1.   This motion
estimator 22 supplies a processed image signal I_p to a
two-dimensional lookup table 14'.

The motion estimator 22 calculates the
correlation coefficients (i.e., normalized cross-
correlation at the zeroth lag) between two consecutive
frames of the image signal I_D.   Preferably, each frame
is divided into subframes, each having a size of 20λ by
20λ, where λ is the wavelength of the center frequency
of the imaging probe.   Such sub-frames reduce
computational requirements, while maintaining valid
motion estimation.   The loss in detail information is
not crucial, since it is still provided in the high
resolution image signal I_D.

The correlation coefficients are then
interpolated and thresholded to distinguish between
different velocities of motion.   Low-pass filtering is
used to provide smooth transitions between regions with
different velocities.   The resulting signal I_p is a
pixel by pixel signal that is used as an addressing

signal for the lookup table 14'. The table 14' can be
chosen to suppress regions with higher velocities of
motion (such as blood pool), and thereby to enhance
tissue contrast. The output signal I_O generated by the
table 14' can encode either a multicolor image or a
monochromatic image as discussed above.

One of the standard operating modes of
ultrasound imaging systems is the spectral Doppler
mode. In this mode, the variation in the power
spectrum of a Doppler shift frequency fd from a sample
volume is displayed as a function of time. Time is
plotted along the horizontal axis, frequency along the
vertical axis. (Magnitude of the spectrum modulates
the intensity.) The Doppler spectrum is proportional
to the velocity distribution within the sample volume.
The spectra from a flat velocity profile have a narrow
bandwidth, while those corresponding to parabolic
profiles have a wider bandwidth. Turbulent flow causes
spectra to fluctuate rapidly with time.

The method described above for the B-Mode
image enhancements can be used to improve spectral
Doppler mode images as well. In this example, the
first image signal I_D is the conventional spectral
Doppler signal $I_D(fd,t)$, and the second image signal I_P
is a processed spectral Doppler signal $I_P(fd,t)$, where

$$I_P(fd,t) = I_D(fd,t) * h(fd) * h(t). \qquad (8)$$

In Equation (8) * signifies convolution, and h(fd) and
h(t) are filters in frequency and time, respectively.
The filters h(fd) and h(t) are selected based on the
feature that is to be enhanced.

For example, to enhance slow variations of
power spectrum magnitude and to reduce noise (i.e., to
increase contrast of the image), in frequency or time,
the filters h(fd) or h(t) are set as low-pass filters.
I_P is then used to color code the magnitude using
Equations (2) or (7). Alternately, to enhance fast
variations, the filters h(fd) and h(t) can be set to be

high-pass filters. For example, by using a high-pass filter h(t) to produce I_P, and by color coding the output as described in Equation (2), turbulent flow or jets can be highlighted. The processor that produces I_P can also be a nonlinear one (or combination of linear filters and nonlinear decision making) in order to extract special features, such as fast, time varying, and high velocity flow, or variance in the estimate.

In this embodiment, the image signal $I_D(fd,t)$ has a wider bandwidth than the image signal $I_P(fd,t)$, and the entity depicted by the image signals is a spectral Doppler plot rather than living tissue. The resulting output signal has a wider bandwidth portion that varies in accordance with $I_D(fd,t)$ and a narrower bandwidth portion that varies in accordance with $I_P(fd,t)$.

From the foregoing, it should be apparent that a wide range of changes and modifications can be made to the embodiments described above. $F(I_P, I_D)$, $F'(I_P, I_D)$ and $\alpha(I_P, I_D)$ can all be adapted as appropriate for the particular application. The display generator may directly determine I_O from I_P and I_D (as by calculation for example), and may thereby eliminate the need for a look-up table. The widest variety of techniques can be used to generate I_P and I_D, either by separate beamforming or by generating I_P from I_D.

The present invention is not limited to use with ultrasonic imaging systems. Instead, it can be used with a wide variety of imaging systems, such as scanning electron microscopes for example. In addition, this invention can be used to enhance images of entities other than physical regions or objects. For example, images of entities such as waveforms or spectral Doppler data displays can be enhanced using the techniques discussed above. It is therefore

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intended that this detailed description be regarded as
illustrative rather than limiting. It is the following
claims, including all equivalents, that are intended to
define the scope of this invention.

WE CLAIM:

1.   An imaging system display processor that is
responsive to first and second image signals depicting
a common entity, said first image signal having
a greater detail resolution than said second image
signal, and said second image signal having greater
contrast resolution than said first image signal, said
display processor comprising:

      a first input for receiving the first image
signal;

      a second input for receiving the second image
signal;

      means, coupled to the first and second
inputs, for forming an output signal as a function of
at least the first and second image signals, said
output signal characterized by a brightness and
a color, said brightness displaying the detail
resolution of the first image signal, said color
displaying the contrast resolution of the second image
signal.

2.   An imaging system display processor that is
responsive to first and second image signals depicting
a common entity, said first image signal having
a greater detail resolution than said second image
signal, and said second image signal having greater
contrast resolution than said first image signal, said
display processor comprising:

      a first input for receiving the first image
signal;

      a second input for receiving the second image
signal;

      a look-up table having at least two
dimensions and coupled to the first and second inputs
and responsive to at least the first and second image
signals to generate an output signal, said output

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signal combining both detail resolution of the first
image signal and contrast resolution of the second
image signal.

3.   An imaging system display processor that is
responsive to first and second image signals depicting
a common entity, said first image signal having
a greater detail resolution than said second image
signal, and said second image signal having greater
contrast resolution than said first image signal, said
display processor comprising:
        a first input for receiving the first image
signal;
        a second input for receiving the second image
signal;
        means, coupled to the first and second
inputs, for forming an output signal as a weighted
combination of at least the first and second image
signals, said weighted combination comprising
a weighing factor that varies in accordance with both
of the first and second image signals.

4.   The invention of Claim 2 or 3 wherein the
output signal encodes a monochromatic image.

5.   The invention of Claim 2 or 3 wherein the
output signal encodes a multicolor image.

6.   The invention of Claim 3 wherein the weighing
factor is selected to emphasize the second image signal
when the second image signal is at a low level.

7.   The invention of Claim 3 or 6 wherein the
weighing factor is selected to emphasize the second
image signal when the second image signal is at a high
level and the second image signal is substantially
greater than the first image signal.

8.   The invention of Claim 3 or 6 wherein the weighing factor is selected to emphasize the first image signal when the first and second image signals are substantially equal.

9.   An imaging system display processor that is responsive to first and second image signals depicting a common entity, said first image signal having a wider bandwidth than said second image signal, said display processor comprising:
        a first input for receiving the first image signal;
        a second input for receiving the second image signal;
        means, coupled to the first and second inputs, for forming an output signal as a function of at least the first and second image signals, said output signal characterized by a wider bandwidth portion that varies in accordance with the first image signal and a narrower bandwidth portion that varies in accordance with the second image signal.

10.   The invention of Claim 9 wherein the output signal encodes a multicolor image.

11.   A display processor for an imaging system that generates a first image signal depicting an entity comprising first regions of a tissue and second regions of blood, said display generator comprising:
        means for generating a second image signal depicting the entity, said second image signal comprising reduced detail resolution and improved contrast resolution as compared to the first signal, said second image signal encoding multiple levels of contrast as appropriate to the tissue throughout the first region; and

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means for superimposing at least the first
and second image signals to generate a composite image
signal which combines detail information of the first
image signal and contrast information of the second
image signal.

12.   The invention of Claim 11 wherein the
generating means comprises a two-dimensional filter.

13.   The invention of Claim 11 wherein the
superimposing means comprises a two-dimensional lookup
table.

14.   The invention of Claim 11 wherein the
generating means comprises a motion estimator.

15.   The invention of Claim 11 wherein the first
image signal is indicative of ultrasonic backscatter
amplitude.

16.   A display processor for an imaging system
that generates a first image signal depicting an entity
comprising first regions of a tissue and second regions
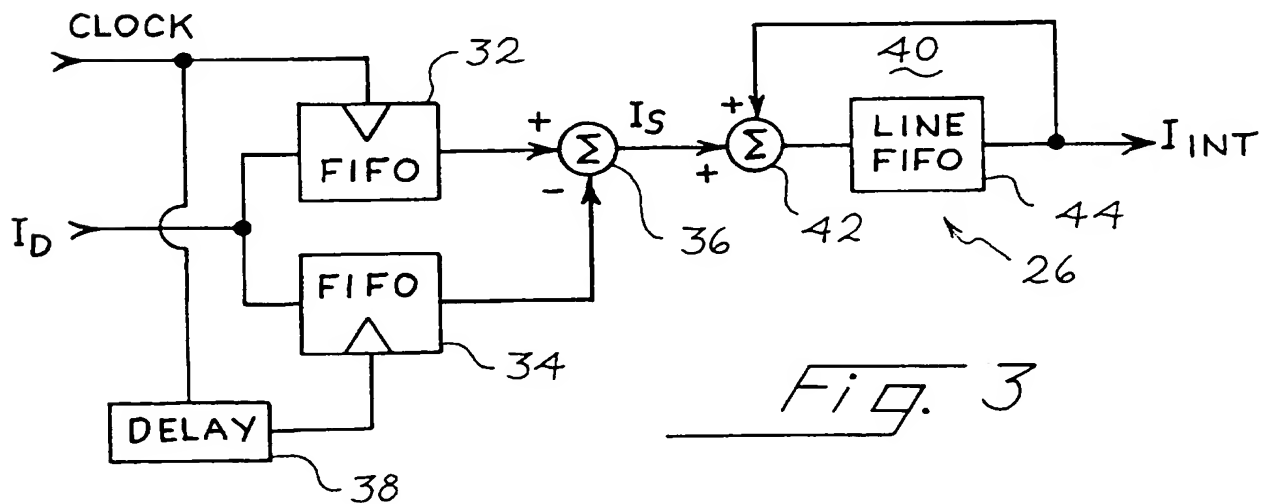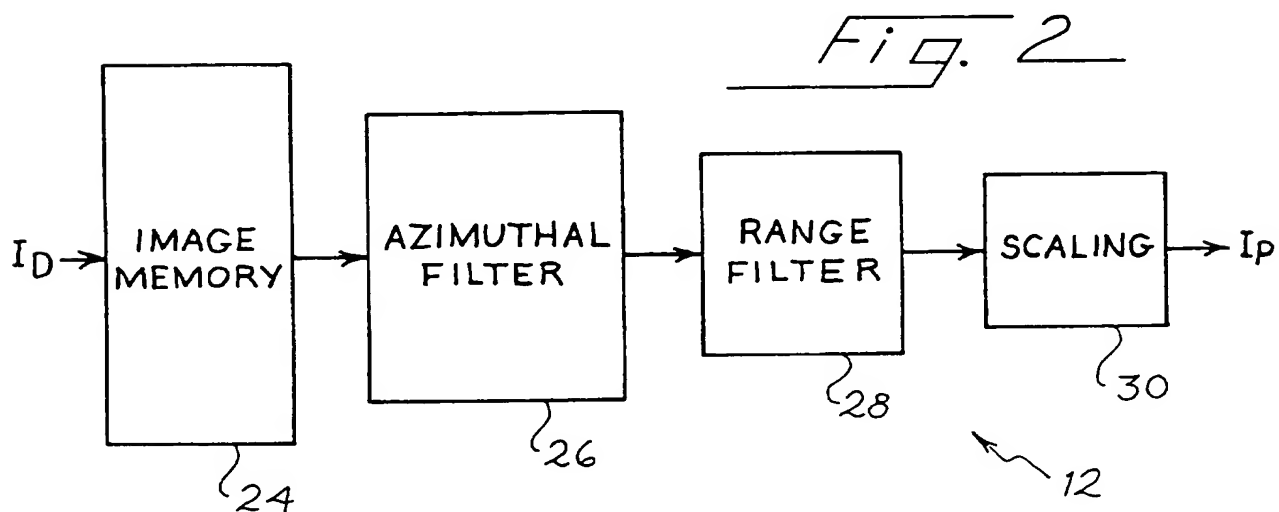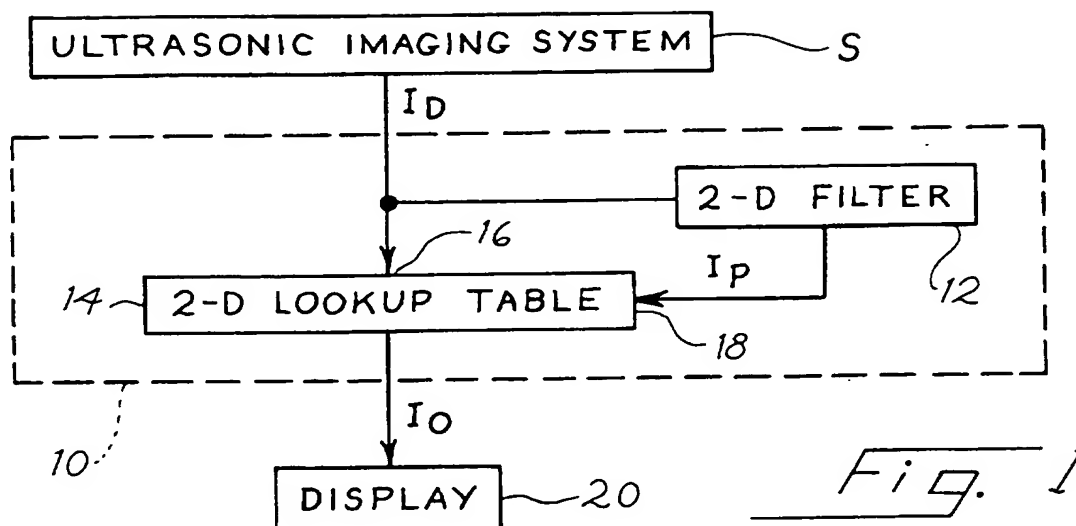of blood, said display generator comprising:
        a motion estimator responsive to the first
image signal and operative to generate a second image
signal, said second image signal comprising reduced
detail resolution and improved contrast resolution as
compared to the first signal; and
        means for superimposing at least the first
and second image signals to generate a composite image
signal which combines detail information of the first
image signal and contrast information of the second
image signal.

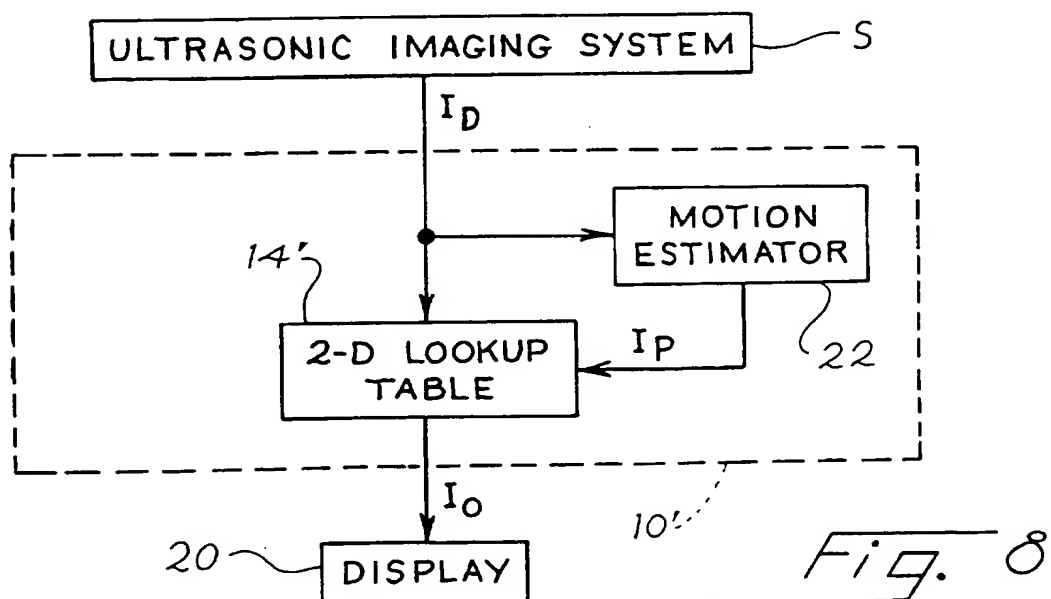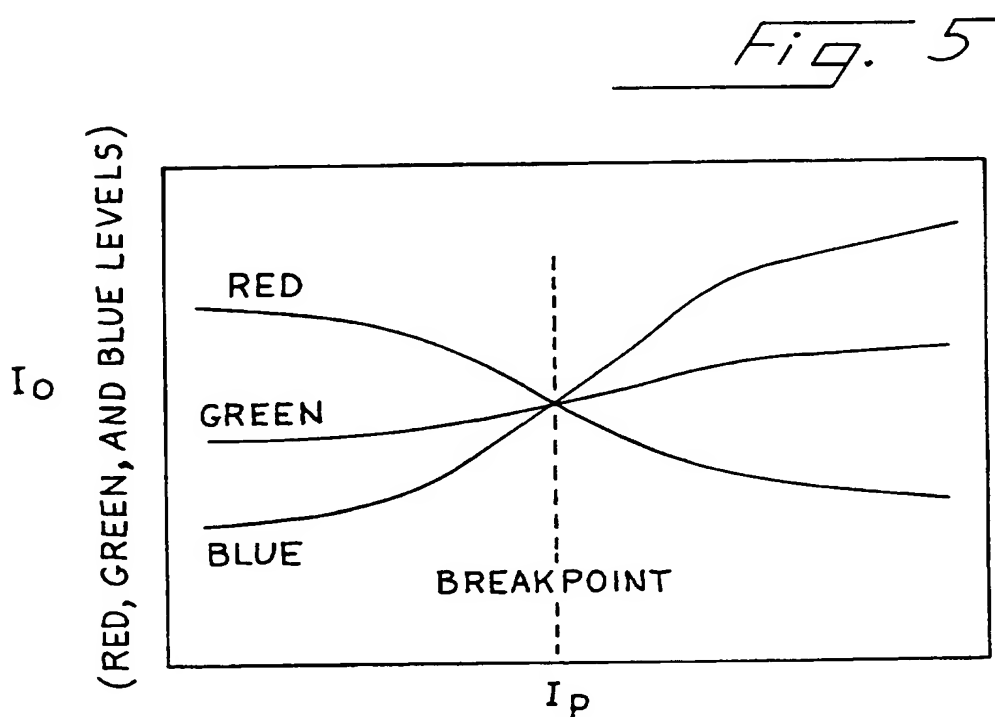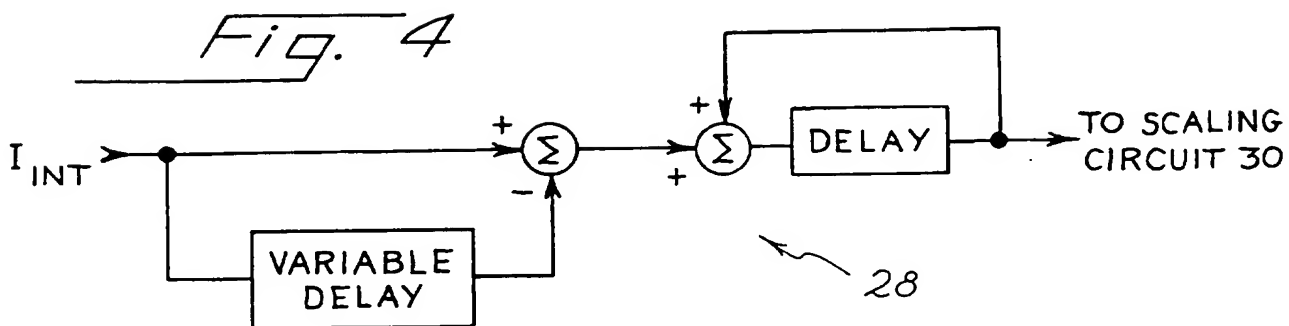17. The invention of Claim 16 wherein the superimposing means comprises a two-dimensional lookup table.

18. The invention of Claim 11 or 16 wherein the composite image signal encodes a multicolor image.
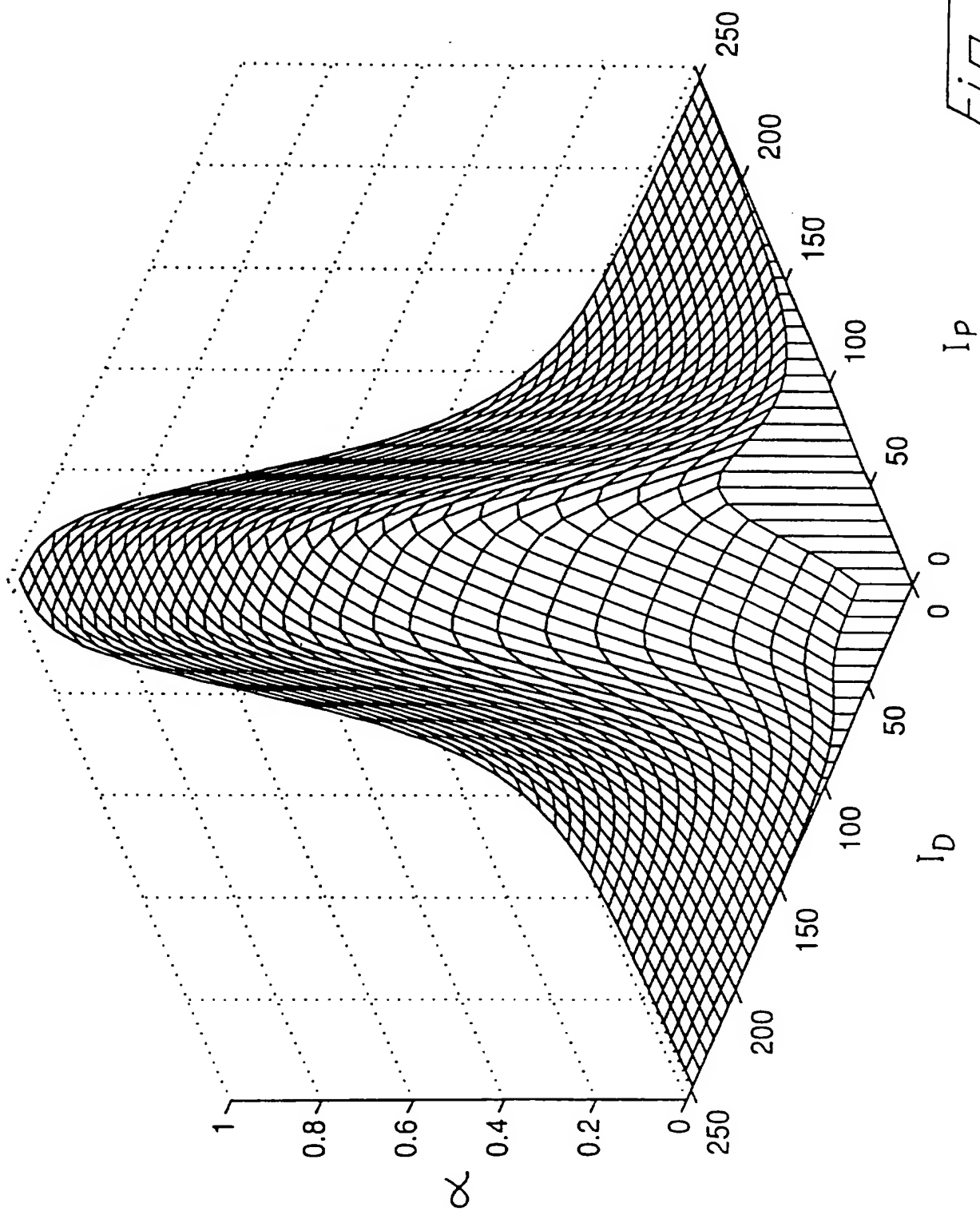
19. The invention of Claim 1 or 2 or 3 or 9 or 11 or 16 wherein the first and second image signals depict ultrasonically measured features of the entity, and wherein the entity corresponds to a physical region.
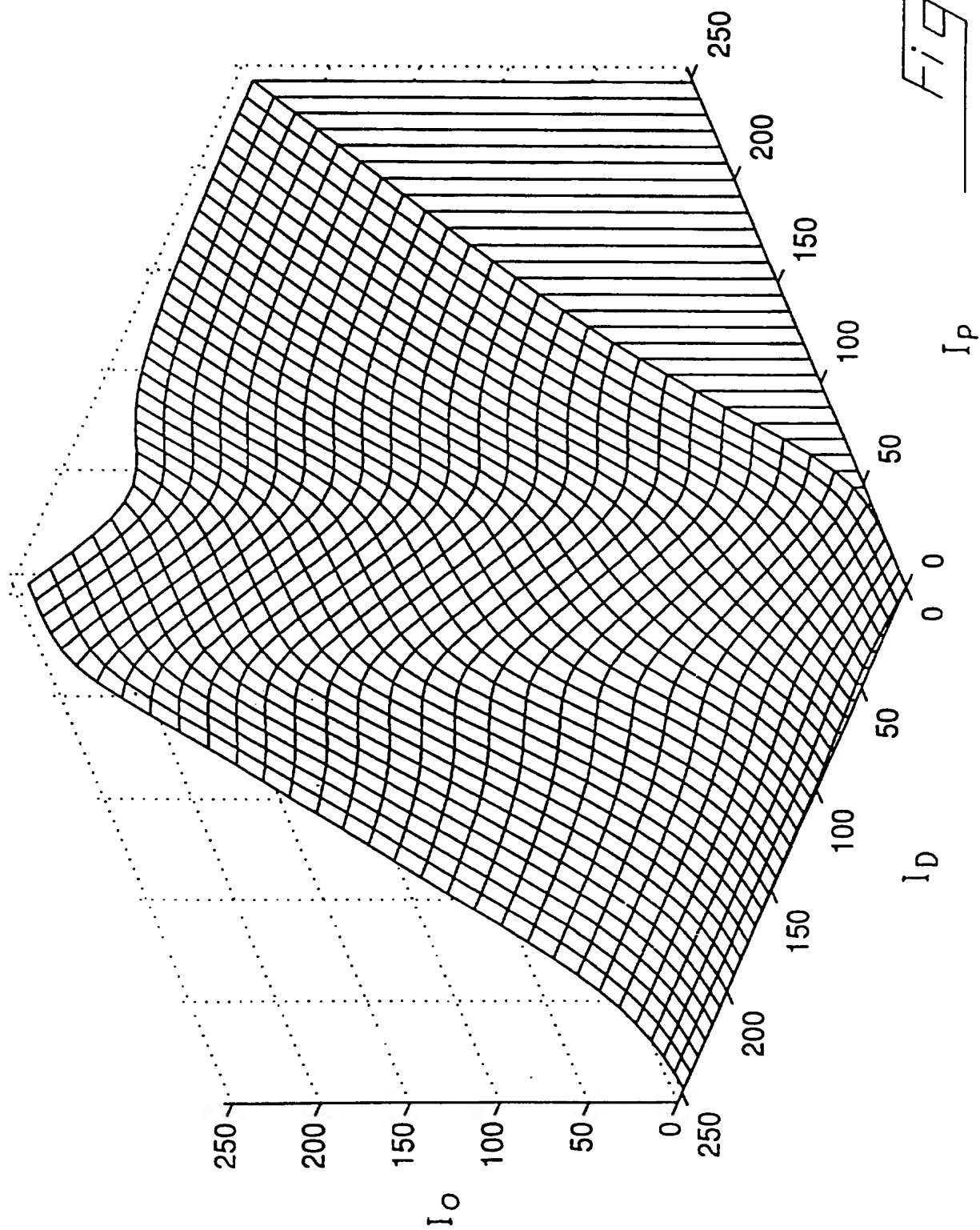
ULTRASONIC IMAGING SYSTEM ~ S

I_D

2-D FILTER

16

14   2-D LOOKUP TABLE

I_P

12

18

10

I_O

DISPLAY ~ 20

Fig. 1

Fig. 2

I_D → IMAGE MEMORY → AZIMUTHAL FILTER → RANGE FILTER → SCALING → I_P

24   26   28   30

12

CLOCK

32

FIFO

I_D

Σ I_S

36

Σ

42

40

LINE FIFO

44

I_{INT}

26

FIFO

34

DELAY

38

Fig. 3

Fig. 4

Fig. 5

Fig. 8

Fig. 6

Fig. 7

INTERNATIONAL SEARCH REPORT

A. CLASSIFICATION OF SUBJECT MATTER

IPC(6) :A61B 8/00
US CL :128/660.04

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

U.S. : 73/602, 625; 128/660.04, 660.05, 660.07; 348/223-225, 242; 358/335

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	US, A, 4,787,393 (FUKUKITA ET AL.) 29 November 1988, see entire document.	1-19
A	US, A, 5,090,411 (HIGUCHI) 25 February 1992, see entire document.	1-19
A	US, A, 5,264,944 (TAKEMURA) 23 November 1993, see entire document.	1-19

☐ Further documents are listed in the continuation of Box C.   ☐ See patent family annex.

*	Special categories of cited documents:	"T"	later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention
"A"	document defining the general state of the art which is not considered to be part of particular relevance		
"E"	earlier document published on or after the international filing date	"X"	document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone
"L"	document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)	"Y"	document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art
"O"	document referring to an oral disclosure, use, exhibition or other means		
"P"	document published prior to the international filing date but later than the priority date claimed	"&"	document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
20 MAY 1996	07 JUN 1996
Name and mailing address of the ISA/US
Commissioner of Patents and Trademarks
Box PCT
Washington, D.C. 20231 | Authorized officer

GEORGE MANUEL |
| Facsimile No.   (703) 305-3230 | Telephone No.   (703) 308-2118 |

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